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(54) Imaging devices systems and methods

(57) An imaging device comprises a semiconductor substrate including an array of pixel cells 18. Each pixel cell comprises an individually addressable active charge storage circuit 20 integral to the semiconductor substrate 16 for accumulating charge resulting from radiation IR incident on the pixel cell. Control electronics can include addressing logic for addressing individual pixel cell active charge storage circuits for reading accumulated charge from the active charge storage circuit of a selected pixel cell. Imaging optimisation can be achieved by determining maximum and minimum charge values for pixels for display, assigning extreme gr y scale or colour values to the maximum and minimum charge values and allocating grey scale or colour values to an individual pixel according to a sliding scale, between the extreme values. Scattered radiation can be detected and discarded by comparing the detected pixel value to a threshold value related to a minimum detected charge value expected for directly incident radiation and discarding detected pixel values less than said threshold value. The device may be used in medical X-ray imaging.

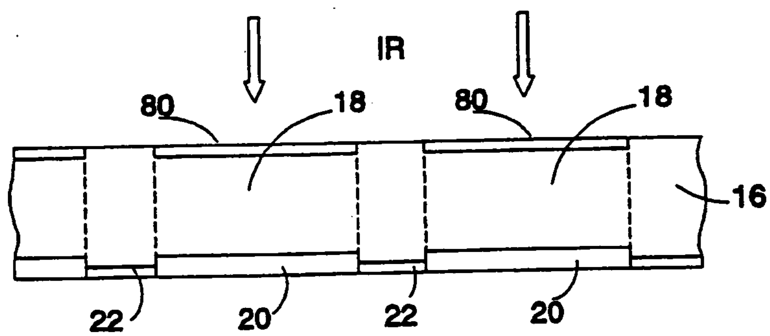

FIG. 5

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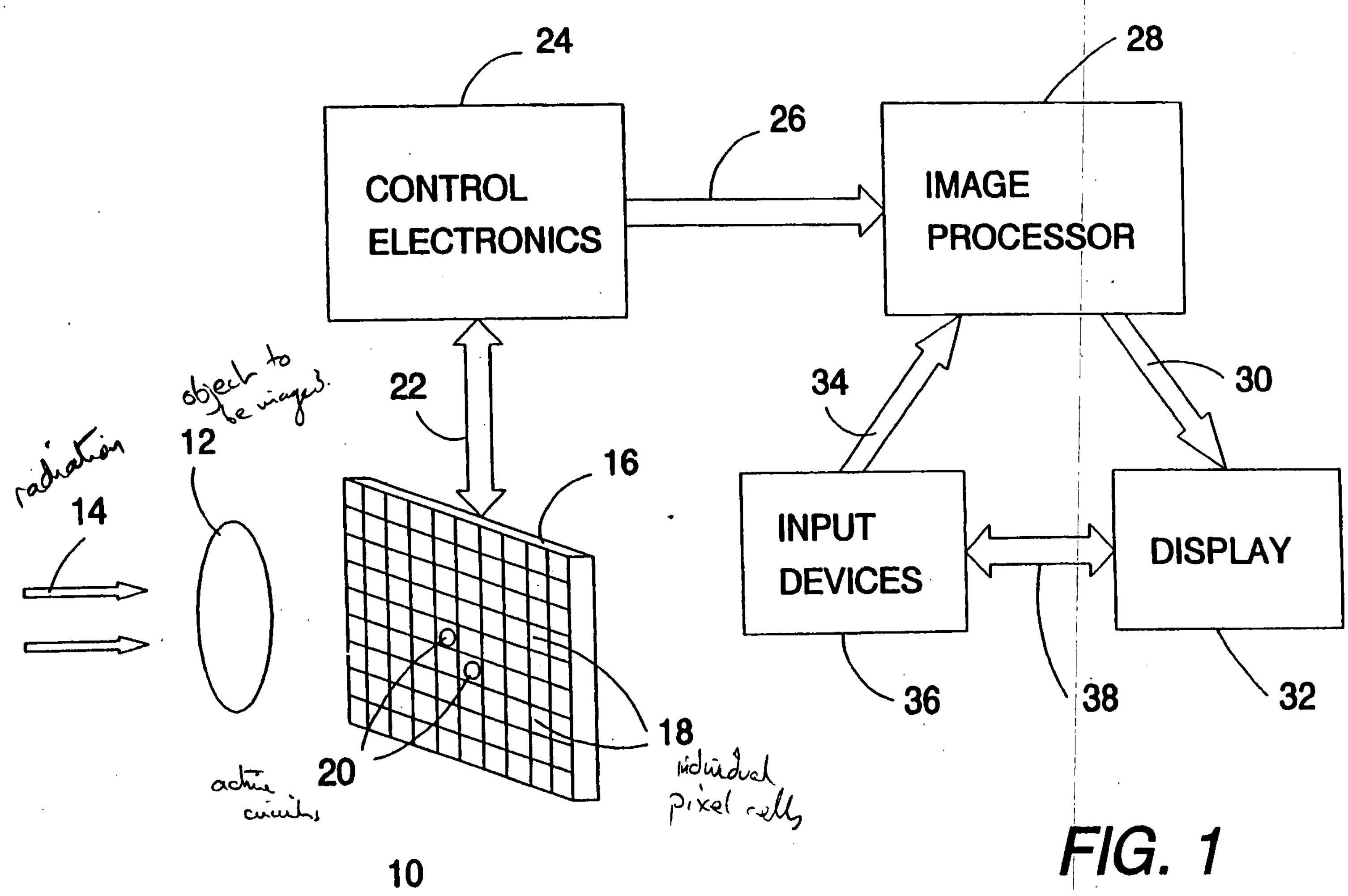
At least one drawing riginally filed was informal and the print reproduced here is taken from a later filed formal copy.



24
CONTROL
ELECTRONICS
26
28
IMAGE
PROCESSOR
22
34
30
radiation
14
object to be imaged.
12
16
INPUT
DEVICES
DISPLAY
20
active circuits
18
individual pixel cells
36
38
32
10

FIG. 1

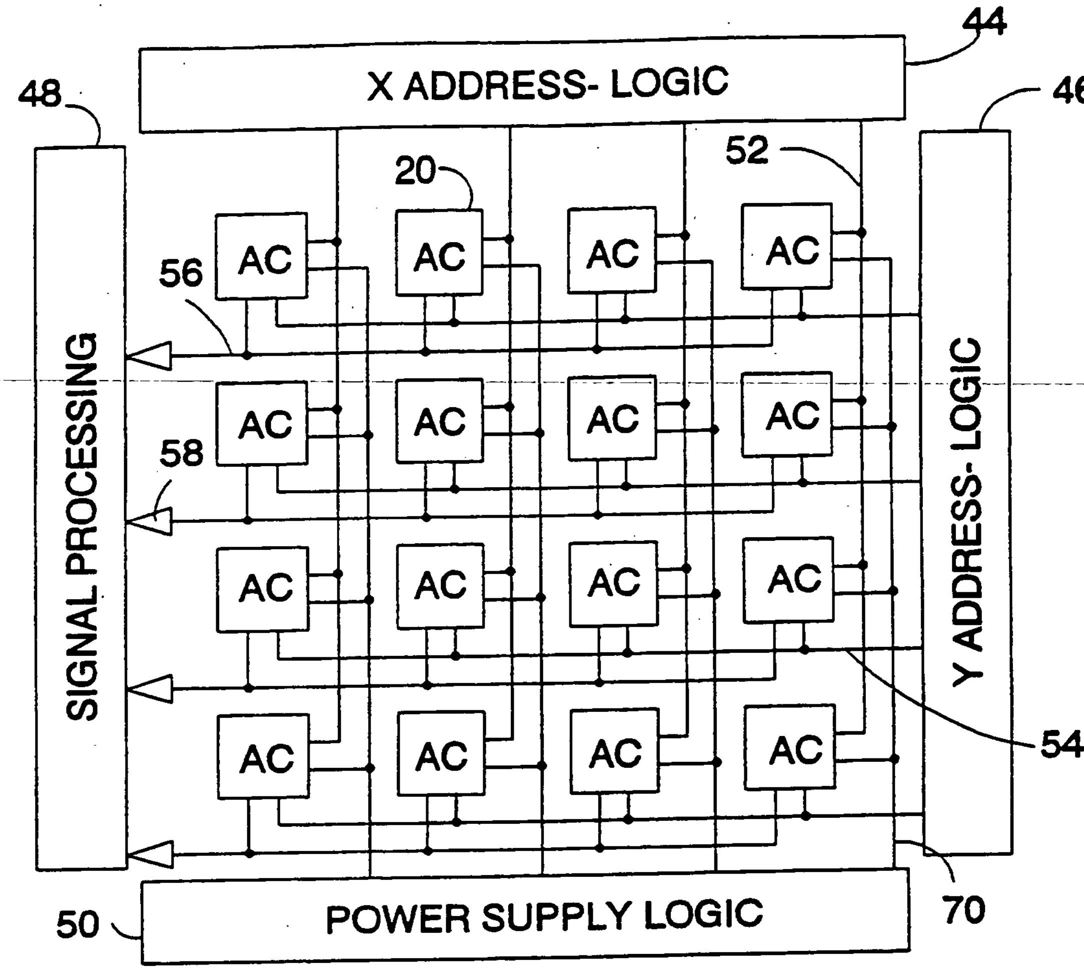

X ADDRESS- LOGIC
44
48
46
52
20
56
58
54
70
50
AC
SIGNAL PROCESSING
Y ADDRESS- LOGIC
POWER SUPPLY LOGIC

FIG. 2

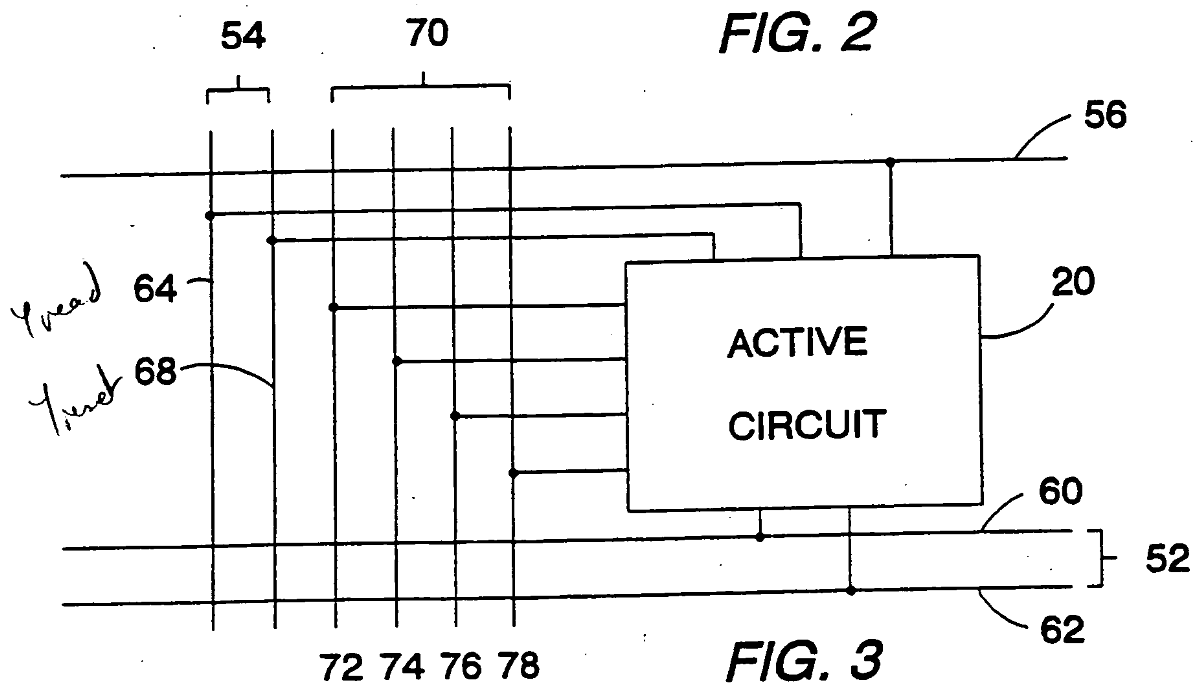

54
70
56
20
Yread 64
Yreset 68
ACTIVE
CIRCUIT
60
52
62
72 74 76 78

FIG. 3

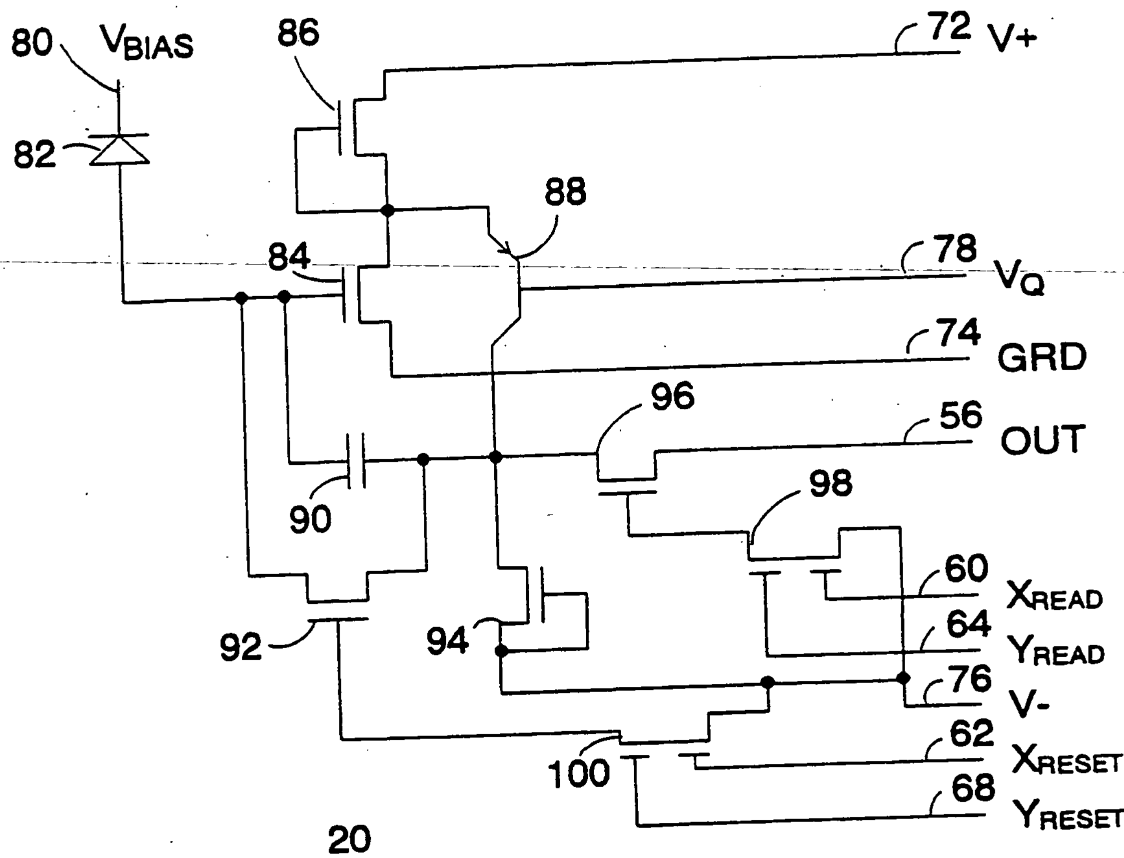

80
V_{BIAS}
82
86
84
88
90
92
94
96
98
100
72
V+
78
V_Q
74
GRD
56
OUT
60
X_{READ}
64
Y_{READ}
76
V-
62
X_{RESET}
68
Y_{RESET}
20

FIG. 4

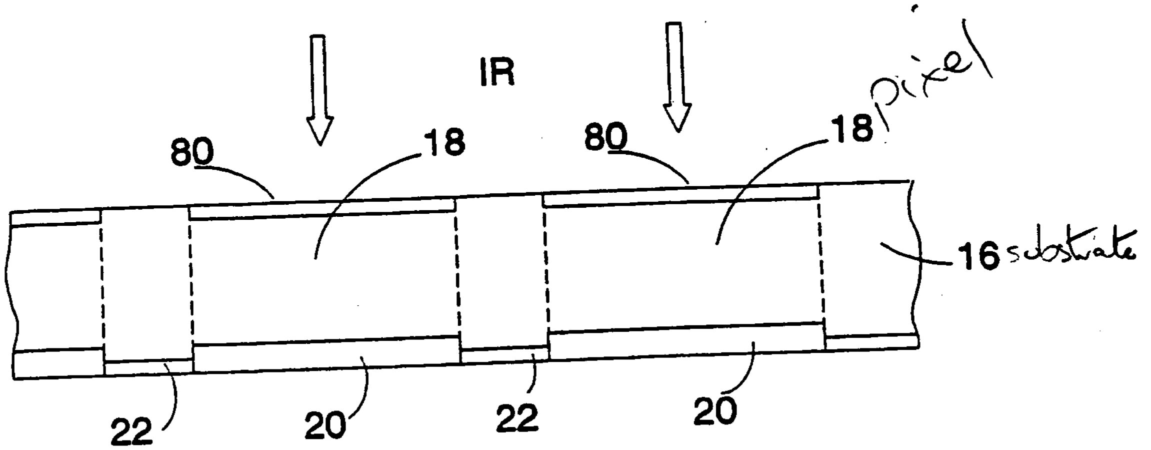

IR
80
18
80
18 pixel
16 substrate
22
20
22
20

FIG. 5

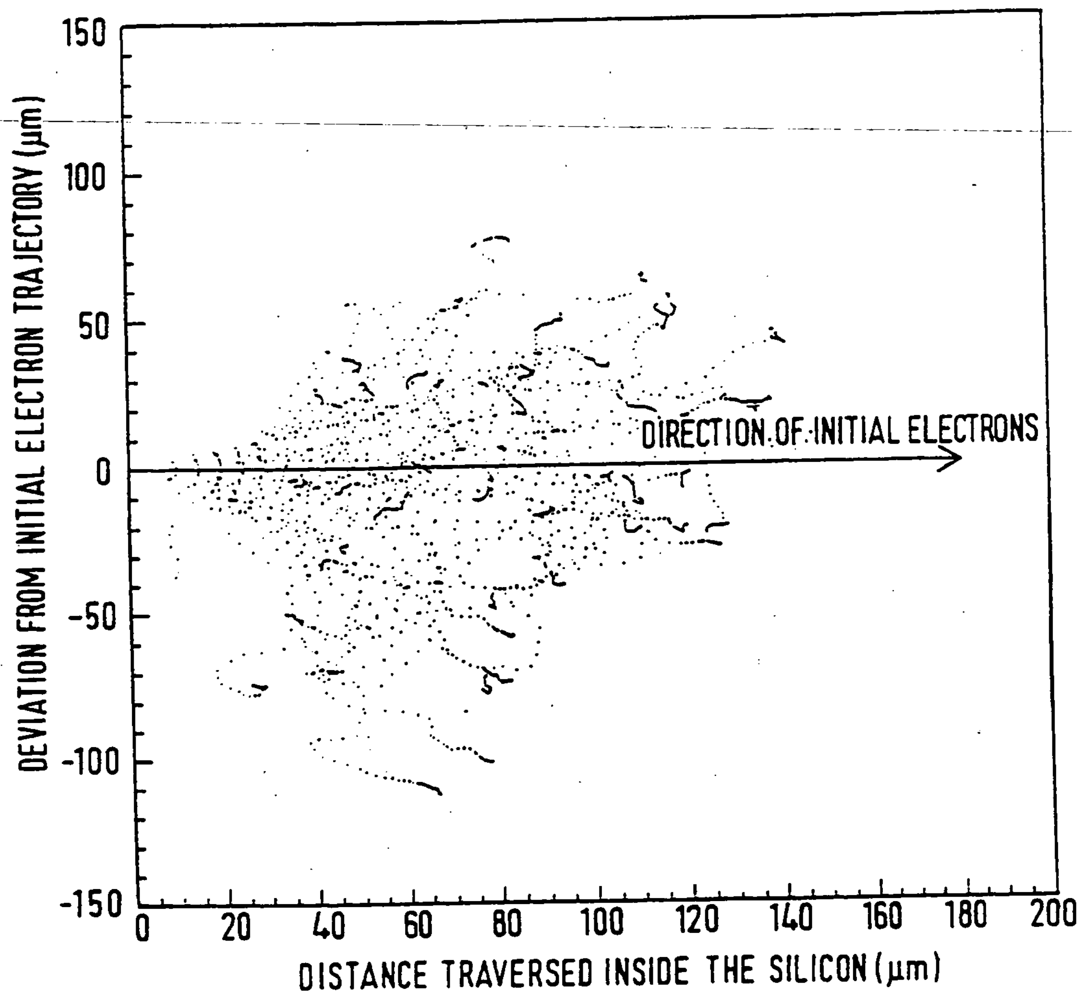

150
100
50
0
-50
-100
-150
DEVIATION FROM INITIAL ELECTRON TRAJECTORY (μm)
DIRECTION OF INITIAL ELECTRONS
0 20 40 60 80 100 120 140 160 180 200
DISTANCE TRAVERSED INSIDE THE SILICON (μm)

FIG. 6

IMAGING DEVICES, SYSTEMS AND METHODS

The invention relates to imaging devices, systems and methods, and in particular to a semiconductor imaging device for use as an image sensor and to an image processing system.

The performance of an imaging system can be assessed, for example, in terms of:

- image resolution, which can be defined by the pixel resolution;
- imaging efficiency, which can be determined in terms of a dose of radiation needed or the exposure time for generating an image;
- dynamic range, which is related to the ability of an imaging device to record different amounts of radiation of different points without saturation and without destroying the single point resolution;
- the response time or image readout speed;
- the analysis, interpretation and storage of images; and
- the imaging area, that is the area of an imaging plane over which an image is collected.

Many different types of imaging devices and systems are known which have differing degrees of performance measured using the parameters listed above for use in differing applications. For example, imaging systems find wide application in medical fields.

Photographic media provide probably the most widely used imaging technique. Photographic media are still widely used in medical applications. Often a photographic medium (e.g. film) is used with a wavelength shifter to increase efficiency, or with digitizers to improve resolution. The photographic film medium does, however, present a number of serious limitations:

- low efficiency;
- poor dynamic range and non-linear response to radiation which can lead to an effective reduction in available resolution;
- passive image recording which limits direct analysis to qualitative analysis (although quantitative analysis can be achieved on digitization of the film image).

Conventional photosensitive devices such as photomultipliers and microchannel plates can provide moderate efficiency and resolution (at the millimetre scale).

Also, wire gas chambers can be used for real time imaging with

electronic recording of the images. These chambers have excellent efficiency and dynamic range but the resolution is at the 500 μm level. In addition it is difficult to provide large imaging areas with such chambers because of the presence of the gas which requires continuous pressure and temperature regulation.

Semiconductor components such as charged coupled devices (CCDs) have been used for imaging purposes as well. Surface channel and buried channel devices are the two most commonly used CCDs. A CCD is typically arranged as an array of pixel detector cells, each of which acts as a pixel storage well. Electrodes define a pixel cell and the voltage applied across the pixel cell creates a natural potential well that can store charge, which remains in the semiconductor substrate until being output.

The operation of a CCD can be summarised as follows. In a first phase, during irradiation, a CCD is in a static state and accumulates charge. In a second phase, the charge is clocked through individual pixel cells and eventually is read out. The transfer of the stored charge from one pixel cell to the next is done by clocking voltages on the pixel cells so that the charge passes from cell to cell in the substrate. The storage capacity of the pixel cells is limited by the capacity of the natural well that is created inside the substrate by applying voltages. Circuitry is provided at the periphery of the CCD and collects the charge after it has been clocked out. Typically, the electron storage capacity of a CCD device is limited to about 700,000 electrons.

A surface channel CCD is a device where the depletion region is a relatively large percentage of the total CCD volume. Interface traps may degrade the transfer efficiency. A buried channel CCD has a relatively small volume which is sensitive to radiation. Both types of CCD have limited storage capacity as mentioned above.

Both types of CCDs are suitable for imaging, for example, low energy X-rays. In such an application, they are typically used in conjunction with a fluorescent screen that shifts the wavelength to the optical region. This increases efficiency, but at the same time reduces resolution because of light diffusion.

Developments in the last few years on silicon radiation detectors have made it possible to manufacture single and double sided strip

detectors. An example of such a detector is described in UK patent application GB-A-2 265 753. The pitch in these detectors can be quite small thus offering excellent resolution. However for applications with high intensities and/or large imaging areas, for example in mammography, the readout speed would need to be hundreds of MHz. When using such devices, the electronic signals from each strip are read out and the position of the incident radiation is determined. One limitation besides the high readout speed needed is the inability of strip detectors to resolve two incident rays that coincide in time and fall within the pitch of a strip, with the result that it is not always possible unambiguously to resolve a point of incident radiation. As a result of these limitations, the resolution and the efficiency degrades.

Pixel silicon detectors comprise a silicon substrate with electrodes which apply a depletion voltage to each pixel. Simple buffer circuits read out the electric signals when a photon is photo-absorbed or when ionizing radiation crosses the depletion zone of the silicon substrate. Every time a charge is present it will be read out and processed. These pixel detectors have a relatively low readout speed per channel because of the number of readout channels. However, they cannot cope with high intensity applications. They also suffer from the problem that it is not always possible unambiguously to resolve the intersection points for simultaneously incident radiation with a resulting reduction in resolution.

Another form of imaging device has also been proposed for X-ray imaging in which a semiconductor substrate defines a pixel array with part of the readout electronics being built on a separate substrate which overlies and is connected to the pixel array substrate by interconnecting microbumps. A device of this type is described, for example, in European patent application EP-A-0 571 135. This type of device is known as a hybrid pixel array to distinguish it from normal pixel arrays. The ability to built large area hybrid pixel arrays is limited by the complexity of manufacturing and assembling the microbumps for interconnecting the two substrates. A hybrid pixel array can store charge and can provide a readout rate of the order of 1000Hz. However, such a readout rate is too low for some applications like mammography.

Accordingly, an object of the invention is to provide an imaging device which mitigates the problems of the prior art.

In accordance with an aspect of the invention there is provided an imaging device comprising a semiconductor substrate including an array of pixel cells, each pixel cell comprising an individually addressable active charge storage circuit integral to said semiconductor substrate for accumulating charge resulting from radiation incident on said pixel cell.

The invention provides an imaging device which can be described as an Active-Pixel Semiconductor Imaging Device (ASID). Embodiments of an imaging device in accordance with the invention are suitable for use in X-ray, β-ray and α-ray real time imaging. An active semiconductor pixel detector is able actively to store charge for individual pixels during irradiation. By enabling each pixel to be addressed individually, the stored charge can be read out at any time during or after irradiation and the readout speed can be optimized to match the radiation intensity.

Accordingly, an imaging device in accordance with the invention can combine the charge storing abilities of a conventional CCD, with the efficiency and speed of a conventional semiconductor pixel detector (which does not store charge). Whereas a normal CCD stores charge in a natural potential well, in an ASID, charge is stored in active electronic structures which are built on the pixels cells of the pixel detector substrate.

Preferably, the active charge storage circuit comprises an integrated charge storage device (e.g., a capacitor, a transistor, a combination of these, etc) for accumulating charge.

Preferably also, each pixel cell active charge storage circuit comprises circuitry for selectively resetting said charge storage device, for example after readout of any charge stored thereon.

The pixel cell size can be approximately 150μm across or less, preferably approximately 50μm across or less and more preferably approximately 10μm across with a substrate between 200μm and 1mm thick.

An imaging system for the imaging device comprises control electronics for the imaging device includes addressing logic for addressing individual pixel cell active charge storage circuits for reading accumulated charg from the active charge storage circuit of a

selected pixel cell, and preferably an analogue to digital converter for converting charge read from a said active charge storage circuit into a digital charge value.

At least part of said control electronics can be integrated into said semiconductor substrate.

Preferably the imaging system comprising an image processor connected to said control electronics for processing said digital charge values from respective active circuits to form an image for display on a display device.

For optimising the display of captured images, the processor determines maximum and minimum charge values for pixels for display, assigns extreme grey scale or colour values to said maximum and minimum charge values and allocates grey scale or colour values to an individual pixel according to a sliding scale between said extreme values in dependence upon the charge value for said pixel.

The grey scale or colour values are preferably allocated in accordance with the following formula:

$$\text{Grey scale value of pixel } i = \mathrm{Min}_{grey} + \frac{(i_{charge} - \mathrm{Min}_{charge})}{(\mathrm{Max}_{charge} - \mathrm{Min}_{charge})} \times (\mathrm{Max}_{grey} - \mathrm{Min}_{grey})$$

In accordance with another aspect of the invention, there is provided a method for imaging accumulated values corresponding to respective pixel positions within a pixel array, such as, for example, charge values accumulated for respective pixel positions of an imaging device as defined above, said method comprising:

- determining maximum and minimum accumulated values for pixels within an area of the pixel array to be imaged;
- assigning grey scale or colour values at extremes of a grey or colour scale to be imaged to said maximum and minimum accumulated values; and
- assigning grey scale or colour values to said accumulated values for individual pixels scaled in accordance with said extreme values; and
- imaging the assigned grey scale or colour values at respective image pixel positions.

In other words, for each portion of an image captured by an imaging device in accordance with the invention, the charge density of all pixels to be displayed is compared, the points of highest and lowest charge density being assigned a colour value at the two extremes of the grey or colour scale being used. The remainder of the pixels points are given a value from the grey or colour scale according to the charge accumulated in the respective pixels.

In accordance with a further aspect of the invention, there is provided a method of automatically optimising imaging using, for example, an imaging system as defined above for different imaging applications where incident radiation leaves a different electrical signal in a pixel cell of a semiconductor substrate dependent on a semiconductor material or compound used and an energy and a type of incident radiation, the method comprising:

- determining an expected best resolution using a centre of gravity technique;
- determining an expected efficiency as a function of radiation type and energy; and
- determining a pixel size and thickness as a function of a selected radiation type and energy and a selected semiconductor material or compound.

This method can also include a step of automatically selecting an imaging device having the determined pixel size and thickness.

This method enables automatic optimisation of the image processing for different imaging applications where, dependent on the semiconductor material or compound used, incident radiation leaves a different electrical signal related to the energy and type of the incident radiation. In accordance with this method, the expected best resolution is identified using a centre of gravity technique. Similarly an expected efficiency is determined as a function of radiation type and energy. For each ASID semiconductor material or compound a database provides values for the various radiation types and energies, thus allowing an immediate and automatic optimization of design specifications. For example if silicon is to be used and the application is mammography at 15keV-40keV with required efficiency 30% the database can determine automatically the pixel size and thickness.

The invention also provides a method for automatically detecting

and eliminating detected pixel value representative of radiation incident on a pixel cell of an imaging device, for example an imaging device as defined above, said method comprising:

- comparing the detected pixel value to a threshold value related to a minimum detected charge value expected for directly incident radiation; and
- discarding detected pixel values less than said threshold value.

Thus this aspect of the invention enables incident radiation (in particular low intensity radiation) that has been scattered before entering the imaging device to be eliminated before processing. This is done by discriminating the detected radiation according to the energy deposited in the form of electrical signals. Because scattered radiation has lost some of its energy it will not pass the minimum energy cut-off.

Another aspect of the invention also provides a method for performing real time imaging of an organic or inorganic object, said method comprising:

- irradiating the object using a radiation source that produces X-rays, γ-rays, β-rays or α-rays;
- detecting at a semiconductor imaging plane or planes of an imaging device as defined above unabsorbed radiation or radiation that is emitted from selected areas of the object, whereby charge resulting from incident radiation at respective pixel cells of said imaging device is accumulated in respective active circuits of said pixel cells;
- addressing said active circuits of said pixel cell individually for reading out accumulated charge;
- processing said read out charge to provide image pixel data; and
- displaying said image pixel data.

An imaging device or an imaging system as defined above can be used for chest X-rays, for X-ray mammography, for X-ray tomography, for computerized tomography, for γ-ray nuclear radiography, for β-ray imaging using isotopes for DNA, RNA and protein sequencing, hybridization in situ, hybridization of DNA, RNA and protein isolated or integrated and X-ray imaging for product quality control.

An exemplary embodiment of the invention is described hereinafter

by way of example only with reference to the accompanying drawings in which:

Figure 1 is a schematic block diagram of an imaging system including an imaging device in accordance with the invention;

Figure 2 is a schematic representation of part of an imaging array and control electronics for an imaging device in accordance with the invention;

Figure 3 is a schematic diagram of the control line connections for a pixel cell of an imaging device in accordance with the invention;

Figure 4 is a schematic circuit diagram of one example of an active circuit for a pixel cell of an imaging device in accordance with the invention;

Figure 5 is a schematic cross-section view of the substrate of an example of an imaging device in accordance with the invention; and

Figure 6 is a schematic illustration of the passage of β-rays through silicon.

Figure 1 is a schematic representation of an example of an application for an imaging system 10 including an embodiment of an imaging device in accordance with the invention.

This application relates to radiation imaging of an object 12 subjected to radiation 14. The radiation may, for example, be X-ray radiation and the object 12 may, for example, be a part of a human body.

The imaging device comprises an Active-pixel Semiconductor Imaging Device (ASID) comprising a plurality of pixel cells 18 configured on a single semiconductor substrate 16 of, for example, silicon. Each pixel cell 18 is defined on the substrate by electrodes (not shown) which apply a biasing voltage to define a depletion zone for the pixel cell 18. Active circuits 20 in the form of electronic structures are defined on each pixel cell 18 to store charge created in the pixel cell when, for example, a photon or a charged particle of radiation is incident on the depletion zone of the pixel cell 18. An example of an active circuit which can be defined on a pixel cell is described hereinafter with reference to Figure 4. An active circuit defined on a pixel cell can be of the order of a few tens of microns in size and can comprise active circuit elements such as, for example, integrated capacitors and transistors or a combination thereof. The

active circuits 20 are constructed integrally to the semiconductor substrate 16 on the pixel cells 18 as part of the semiconductor wafer fabrication process. Conventional wafer manufacturing techniques can be employed for fabricating the semiconductor wafer.

The pixel cells 18 of the semiconductor substrate 16 are formed with a depletion zone such that, when a photon is photo-absorbed in the semiconductor substrate 16 at a pixel cell 18 creating an electric charge or when a charged radiation ionizes the depletion zone of the semiconductor substrate 16 at a pixel cell 18, an electric pulse flows from the semiconductor substrate depletion zone to the active circuit 20 for that pixel cell 18. The pixel cells 18 thus form radiation detectors. The charge associated with the electric pulse is then stored in an active circuit element such as an integrated capacitor or at the base junction of an integrated transistor such that new charge created from subsequent incoming radiation is added continuously. The charge storing process in an active circuit 20 continues until control signals are issued from control electronics 24 to start a process of reading out information from each individual pixel cell. After the charge readout from a pixel cell has been completed, the pixel cell circuit is active again to accumulate new charge. As each pixel cell stores charge in a respective active circuit which can be addressed individually, the problem of ambiguous points experienced with the prior art can be avoided completely.

The pixel pitch can be as small as 10 µm which results in excellent position resolution and consequently excellent image resolution.

The control electronics 24 include processing and control circuitry which is connected to the pixel cells 18 on the semiconductor substrate as represented schematically by the two-way arrow 22. The control electronics 24 enable the active circuits 20 associated with individual pixel cells 18 to be addressed (e.g., scanned) for reading out charge accumulated in the active circuits 20 at the individual pixel cells 18. The charge read out is supplied to Analogue to Digital Converters (ADCs) for digitisation and Data Reduction Processors (DRPs) for processing the binary signal.

The processing which is performed by the DRPs can involve discriminating signals which do not satisfy certain conditions such as

a minimum energy level. This is particularly useful when each readout signal corresponds to a single incident radiation event. If the energy corresponding to the measured signal is less than that to be expected for the radiation used, it can be concluded that the reduced charge value stored results from scattering effects. In such a case the measurement can be discarded with a resulting improvement in image resolution.

The control electronics 24 is further interfaced via a path represented schematically by the arrow 26 to an image processor 28. The image processor 28 includes data storage in which it stores the digital value representative of the charge read from each pixel cell along with the position of the pixel cell 18 concerned. For each pixel cell 18, each charge value read from the pixel cell is added to the charge value already stored for that pixel cell so that a charge value is accumulated. As a result, each image can be stored as a representation of a two-dimensional array of pixel values.

The image data can be stored, for example, in a database as a two-dimensional array for the image:

Image $(1{:}N_{pixels}, 1{:}3)$

where the first index includes N_{pixels} items representing a pixel number on the imaging plane which runs linearly from one to a maximum pixel number N_{pixels} and the second index includes three values, for the x and y coordinates and the charge value accumulated for each pixel, respectively.

The image processor 28 can access the stored image data in the database to select a given image (all the array) or a part of the image (a sub-sample of the image array). The image processor 28 reads th values stored for the selected pixel positions and causes a representation of the data to be displayed on a display 32 via a path represented schematically by the arrow 30. The data can of course be printed rather than, or in addition to being displayed and can be subjected to further processing operations. Background and noise can be subtracted as a constant from each pixel charge value.

Preferably, before displaying, printing or further processing the image data, the image processor 28 finds the two extreme pixel charge values stored for the pixels selected and assigns thes values to the

two extremes of the grey or colour scale which can be used for displaying, printing or further processing of the image, as appropriate. The remaining charge values for the pixel positions can then be assigned an intermediate grey scale or colour value between these extreme values according to the charge deposited on the pixel. For example the grey scale value can be assigned to the charge values for individual pixels in accordance with the following equation:

$$\text{Grey scale value of pixel } i = \text{Min}_{grey} + \frac{(i_{charge} - \text{Min}_{charge})}{(\text{Max}_{charge} - \text{Min}_{charge})} \times (\text{Max}_{grey} - \text{Min}_{grey})$$

The selection of a portion of the image to be zoomed can be achieved by means of conventional user input devices 36 via a data path represented schematically by the arrow 34, possibly interacting with the display 32 as represented schematically by the double arrow 38. The user input devices 36 can include, for example a keyboard, a mouse, etc.

Figure 2 is a schematic representation of the control electronics 24 in more detail and the relationship of the control electronics 24 to the active circuits 20 of the pixel cells 18 on the substrate 16. For ease of illustration an array of 16 pixel cells is illustrated in Figure 2 and only some of the signal lines which make up the path 22 in Figure 1 are shown. It will be appreciated that an imaging device in accordance with the invention will normally include a significantly larger number of pixel cells 18 than are shown in Figure 2.

The control electronics 24 include X address logic circuits 44, Y address logic circuits 46, power supply circuits 50 and signal processing circuits 48. Preferably some, if not all, of the control electronics 24 is implemented on the substrate 16 at the periphery of the image array formed by the array of pixel cells 18.

The power supply circuits 50 provide power for the individual active circuits 20 on the pixel cells 18 via lines 70, shown schematically on Figure 2 and can additionally be arranged to supply the biasing voltage via lines (not shown) for the electrodes defining the pixel cells.

Th X and Y addressing logic 44 and 46 provide signals via row and column lines 52 and 54, respectively (also shown schematically in

Figure 2), for controlling the reading and resetting of the individual active circuits 20 of the pixel cells 18.

The signal processing circuitry 48 is connected to output lines 56 shown schematically in Figure 2 for the active circuits 20 of the pixel cells 18. As shown in Figure 2, one output line is provided for each row of pixel cells 18 and is connected via an output amplifier 58 to the signal processing circuitry 48. However, it will be appreciated that as alternatives separate output lines could be provided for each column, or for groups of rows or columns or for groups of pixel cells as desired.

Figure 3 illustrates in more detail the lines which are provided between the control circuitry 24 and the active circuit 20 for a pixel cell 18 stage cell in accordance with one embodiment of the invention. In this embodiment the power supply lines 70 comprise a positive supply line V+ 72, a ground line GRD 74, a negative supply line V- 76 and an bias voltage line V_Q 78. The row lines 52 comprise an Xread line 60 and a Xreset line 62 and the column lines 54 comprise a Yread line 64 and a Yreset line 68. One output line 56 is provided for each row in this embodiment as has already been explained.

Figure 4 is a circuit diagram of an example of an active circuit 20 for a pixel cell 18 in accordance with one embodiment of the invention.

The pixel cell 20 is represented by the diode symbol 82 connected to the voltage bias Vbias 80, this being applied via the electrode (not shown) defining the depletion volume of the pixel cell 18.

Charge created by radiation incident on the depletion volume of the pixel cell 18 is input to the base of a first, input transistor 84 (here a field effect transistor (FET) having a transconductance of, for example, 0.3mS and a drain source current value I_{DS} of 100μA and a capacitance of 0.1pF). The source and drain of the input FET 84, are connected between a first current source 86 (here a suitably configured FET, although this could be replaced by a resistor) and the ground line GND 74 shown in Figure 3. The current source 86 is in turn connected to the positive supply line V+ 72 shown in Figure 3.

The junction between the input FET 84 and the current source 86 is connected to one terminal of a second transistor 88 forming a common

base bipolar amplifier controlled by the bias voltage applied to its base. The base of the second transistor 88 is connected to the bias voltage line V_Q 78. The remaining terminal of the second transistor is connected via a feedback capacitor Cf 90 (e.g., with a capacitance of 0.3pF) to the base of the input FET 84.

The junction between the second transistor 88 and the capacitor Cf 90 is also connected to a second current source (here a suitably configured FET, although this could be replaced by a resistor) to the negative supply line V- 76 illustrated in Figure 3. Charge resulting from radiation incident on the depletion volume of the pixel cell can thus be accumulated at the capacitor Cf 90.

The X and Y read lines, Xread 60 and Yread 64, are connected to read logic 98 (here a dual base FET) which in turn is connected between the negative supply line V- 76 and an output switch 96 (here a FET) whereby charge collected on the capacitor Cf 90 can be output via the output line 56 when a signal is supplied on the Xread and Yread lines 60 and 64 simultaneously. The X and Y reset lines, Xreset 62 and Yreset 68, are connected to discharge logic 100 (here a dual base FET) which in turn is connected between the negative supply line V- 76 and a discharge switch 92 (here a FET 92) for discharging and thereby resetting the capacitor Cf 90 when a signal is supplied on the Xreset and Yreset lines 62 and 68 simultaneously.

The circuit shown in Figure 4 forms a charge sensitive amplifier with charge storage capability in the feedback capacitor Cf 90 and with output and resetting circuitry. Depending on the charge storage time and radiation hardness requirements, the FETs can be implemented by an appropriate technology such as JFET or MOSFET. If the capacitor Cf 90 has a capacitance of 0.3pF, this corresponds to a storage capacity of about 1.8 million electrons. If the capacitor Cf 90 has a capacitance of 1pF, this corresponds to a storage capacity of about 6 million electrons. The maximum output clock frequency with a reset FET in the output line is 1 - 5 MHz. This maximum output frequency reduces to about 100kHz without a reset FET in the output line.

The circuitry illustrated in Figure 4 could be implemented on, for example, a pixel cell having a size of approximately 150 x 150 microns. Depending on the circuit technology used, small r pixel siz s are also envisaged. For example, the active circuit 20 could also be

implemented for pixel cells having a size of the order of 50 x 50 microns. A total of 1200 x 1200 pixel cells could be implemented on a single semiconductor substrate 16, giving an imaging area of the semiconductor substrate of approximately 60mm by 60mm. Multiple such planes can be placed next to one another thus giving, for example, an imaging surface as large, or larger than 400mm x 400mm. Around the outside of the imaging area formed by the array of imaging cells some or all of the control electronics 24 may also be implemented as an integral part of the semiconductor substrate wafer 16.

The active circuit shown in Figure 4 along with the connections shown in Figure 3 are implemented integrally on the semiconductor substrate using conventional integrated circuit manufacturing techniques.

Figure 5 is a very schematic cross-sectional view of the semiconductor substrate 16 of an example of an imaging device in accordance with the invention. It is assumed in Figure 5 that incident radiation IR will be incident in a downwards direction onto the upper surface of the substrate 16 as represented in Figure 5. Each pixel cell 18 is defined by an electrode to which is supplied a bias voltage at 80. The extent of the pixel cell 18 is represented schematically by the dotted lines. For each pixel cell, active circuit elements 20 are provided towards the rear (here the bottom) surface of the semiconductor substrate. The active circuits 20 are connected to one another by means of the paths 22. The circuitry is provided to the rear of the substrate in order that it does not reduce the degree of radiation which can be incident on the pixel cell. It will be appreciated the Figure 5 is merely a schematic representation on the semiconductor substrate and is not shown to scale.

It will be appreciated that the size of the pixel cells and the number of pixel cells which can be implemented on a single semiconductor detector will depend on the particular semiconductor integration technology used. Thus, although particular examples of sizes and component values have been given, the invention is not limited thereto and is intended to include changes in those dimensions and values as are possible with current such technology and will be possible with future technology. Also, it will be appreciated that the actual circuits shown, for example the active circuit 20 shown in

Figure 4 and the connection lines and control circuitry illustrated in Figures 3 and 2, respectively, are merely examples of possible circuits and that many modifications and additions are possible within the scope of the invention.

The invention brings a number of advantages as a result of accumulating charge in an active circuit on a pixel cell.

The ability to store the charge in the active circuits on the pixel cells and then selectively to read out the stored charge from individually addressable active circuits in one-to-one correspondence with the pixel cells completely resolves any ambiguities regarding the point of incidence of concurrently incident radiation.

As the charge can be built up over a period on individual active circuits, the readout speed need not be excessively high, with the result that, for example, software-based generation and processing of the image in real time is possible and indeed can be implemented inexpensively on readily available computer hardware.

For each portion of the captured image the contrast and resolution can be adjusted automatically and displayed on a full screen. Wherever there is a charge density variation between the pixel cells of an area of the image captured by the imaging device, features of the image can be resolved when that part of the captured image is displayed.

The dynamic range is effectively unlimited assuming that the charge from the charge storage device of the pixel cell active circuits is read and the charge storage device is reset repeatedly before the storage capacity of the charge storage device is exhausted. It is merely necessary to select the "refresh rate" of the active circuits, that is the frequency of reading out and resetting those circuits, to suit the storage capacity of the charge storage devices and the anticipated maximum radiation density. Thus, as more radiation creates more charge, this is stored in the active circuits of the pixel cells, then read out at appropriate intervals and digitized by the control electronics. After digitization, the charge has a known value that can be accumulated with existing digitized charge values of the same pixel. The only practical limitation is the maximum digital value which can be stored by the processing circuitry. However, even then the processing circuity could be arranged to detect a value approaching the maximum

possible value which can be stored and then to apply a scaling factor to the stored values of all pixel cells.

The invention enables real-time imaging. Once an image array has been created, even before irradiation starts, the image array can be updated continuously with new digitized charge values from the imaging device, which charge values are then added to the existing charge values of the respective pixel of the array and the accumulated charge values are displayed in real time.

Where a continuously updated image array is employed, this provides an efficient use of computer storage as detected radiation will not yield more image points, as is the case with some prior techniques, but instead yields higher charge values for the pixel cell positions concerned. In other words, the present invention enables the accumulation of radiation counts rather then generation of an ever increasing number of radiation hit points.

Thus, whereas CCDs suffer from relatively low sensitivity and storage capacity and are static devices, and conventional semiconductor pixel detectors are more sensitive with high resolution but require high readout speeds and do not store charge, an imaging device of the ASID type in accordance with the invention can provide the advantages of conventional semiconductor devices with the additional advantages of enabling charge storage at the pixel cell level with random access readout at a lower rate. Embodiments of the present invention can provide the storage of electric charge on each pixel for up to, for example, 25,000,000 electrons (approximately 36 times more than CCDs). The pixel thickness that is fully depleted can be up to 1mm thus making an imaging device in accordance with the invention very sensitive to X-rays with energies less than 200keV. For charged radiation the sensitivity can be practically 100%. The minimum pixel thickness can be only 200 μm which gives improved resolution when low energy charged radiation is detected. With this in mind the dead layer of the semiconductor substrate which is insensitive can be as thin as 0.5μm so that the signal of β-rays with energy less than 30keV is not lost.

To illustrate the improvements the invention can bring, an example of an application of the invention to mammography will be described.

Typically the radiation dose in mammography is of the order of

10^{15} photons in one sec. If the surface of the imaging plane is 300mm x 300mm and the pixel size is 100µm then there are 100 x 10^6 photons per pixel in one second. This means that the readout speed per channel must be at least 100MHz which is technically and economically unrealistic. By way of example silicon can be used as the semiconducting material in our invention. Then every 3.6eV will yield an electron-hole pair. Assuming that 25 x 10^6 electrons can be stored in an imaging device in accordance with the invention, the signals from 4500 photons with an energy 20keV each can be accumulated before the stored charge must be read out. The readout speed need only be 22 kHz in this example, rather than the 100MHz which would be required using semiconductor pixel detectors in accordance with the prior art.

For applications with radiation intensities requiring less than the maximum achievable readout speed per pixel (a few MHz), the present invention offers a way to discriminate against scattered radiation which, if not excluded, will degrade the image resolution. The way that this can be done will now be explained. The charge created from each and every photon or charged radiation particle is first stored in the active circuits of the pixel cells and then read out. The control electronics digitises the charge and the DRP can compare the digitized value to a threshold reference value. The reference value corresponds to the charge to be expected from incident radiation of the type in question, that is for example an X-ray of a given wavelength or from a charged radiation of a given energy. The digitised charge value is then excluded from further consideration if it is less than the reference value. This discrimination operation enables scattered rays to be eliminated from consideration. When inelastic scattering effects occur before the imaging plane while, for example the radiation traverses an object under observation, the scattered radiation loses some of its energy before the imaging plane so that less charge is created in the depletion region of a pixel cell. Such effects are Compton scattering for photons and ionization scattering for charged particles.

Imaging device design optimisation in accordance with the invention can be carried out in an predetermined automated manner. Each material or compound chosen for the semiconductor substrate has a

different response to incident radiation which depends on the physical properties of the material or compound, the radiation type and the radiation energy. A centre of gravity method is applied to the deposited electric signal at every step as incident radiation traverses the semiconductor substrate. This enables the best attainable resolution to be determined as a function of the above parameters. Thus the pixel size is determined. By correctly choosing the pixel size the signal to noise ratio can be maximised (because most of the signal is contained in one pixel) while the cost and device complexity is minimized. These results along with the expected sensitivity can b stored in a database and can be used to define the design parameters of the imaging plane of the imaging device, namely the pixel size and substrate thickness. Alternatively, a series of imaging planes compatible with a common set of control electronics and image processor can be provided. An end user can then, before carrying out imaging, input a desired sensitivity to the image processor to cause this automatically to select an imaging plane with the correct specification.

Consider, as an example, the use of silicon as the semiconductor substrate material. In biotechnology applications, isotopes such as 3H, 35S, 32P, 33P, 14C and 125I are used. These isotopes emit β radiation. Consider 35S, for example, which emits 170keV charged radiation. Figure 6 shows the passage of many such β-rays through silicon. If the centre of gravity method is applied, it is found that the resolution cannot be better than 32μm. The pixel size can then be chosen to be greater than 32μm in order to contain most of the electrical signal. The β radiation isotopes mentioned above are used in most biotechnology applications. In mammography, tomography, nuclear medicine and product quality control X-rays are used with energies between 10keV-180keV.

By way of example if silicon is chosen as the semiconductor material, we present in Table 1 design values for the image plane.

	Pixel Size (μm)	Pixel Thickness (mm)
Isotopes (β-rays) 3H(18keV)	<10	0.2
14C(155keV)	27	0.2
35S(170keV)	32	0.2
33P(250keV)	58	0.2
32P(1700keV)	<10	0.2
α particles For energy>100keV	<10	0.2
Photons X-rays (10keV-30keV)	<10	0.3
X-rays (30keV-100keV)	<10	≥1.0
X-rays (>100keV)	50	≥1.0

Table 1: Design specifications as a function of radiation type and energy. These values enter a database and before execution of an application automatically determine the optimized imaging plane design.

The preferred embodiments of the current invention include imaging in biotechnology and X-ray imaging in mammography and tomography, γ-ray cameras (nuclear medicine) and automatic product quality control.

There are many biology applications that perform imaging with β radiation. Most often one of the following isotopes are used: 3H(18keV), 14C(155keV),35S(170keV), 33P250keV), 32P(1700keV).

The precision requirements for these applications could be summarized as follows:

- hybridization in situ requires ideally 10 μm;
- hybridization on DNA, RNA and protein isolated or integrated requires ideally better than 300 μm;
- Sequences of DNA require ideally 100 μm.

An imaging device in accordance with the invention can meet the above requirements. In addition the excellent efficiency (practically 100%) of imaging devices in accordance with the invention can reduce th time for obtaining the results from days or months to hours. Since the imaging is done in real time a biologist can see the results while

they are being accumulated. Software and statistical methods of analysis can be used for interpreting these results.

In mammography the X-rays used have typically energy from 10keV to 30keV. The X-ray source is placed behind the object under observation which absorbs part of the X-rays and lets the rest through. The X-rays that arrive at the imaging plane are consequently photo-absorbed and create an electrical signal from which the point of incidence is determined. The charge density distribution effectively defines the image, which, with on-line conventional processing can be coloured, zoomed and analyzed with maximum image contrast and resolution. With 0.3mm thick active silicon pixels (or more) the efficiency is very good and the dose needed can be reduced drastically. The resolution for mammography can be better than 20μm and organic structures of that size are revealed.

In nuclear medical diagnosis an isotope emitting X-rays at the range of 150keV (such as, for example, Tc^{99} with 6 hours half life) is injected to the human body and concentrates to certain areas that are imaged. The radiation is emitted isotropically and around the human body collimators filter away unwanted directions thus making projections of a point to different planes. According to the current invention the ASID can be placed in front of and around the human brain replacing existing imaging planes.

Yet another possible application of the invention is non-destructive industrial evaluation and product quality control. Depending on the inorganic object that is observed a different X-ray energy is chosen so as to optimize resolution with high contrast and efficiency. X-ray energies in the range 20keV-180keV may be used. The image of a product or a structure is automatically compared to an ideal image of the same product or structure and various levels of severity may trigger different actions that give feedback to the production line.

Thus there has been described a new device and method for imaging in applications that use any type of radiation (X-rays, γ-ray, β-rays and α-rays). An automated system for real time, high resolution, high efficiency and unlimited dynamic range imaging can be provided using the new active pixel semiconductor radiation detectors devices. The active pixels are defined on a single semiconductor substrate and

charge is stored on circuit elements that are built on the pixels. The pixels can be read out individually and all ambiguities that are present in a normal strip or CCD semiconductor detector are resolved. The imaging plane design can be predetermined depending on the radiation type, energy of radiation and object under observation. The stored charge on each pixel can be used to acquire an image with automatically adjusted high contrast and resolution. In one application the invention can be used in X-ray and γ-ray radiography with radiation energies from 10keV to 200keV. In another application the invention can be used for DNA, RNA and protein sequencing, in hybridization in situ and in hybridization on DNA, RNA and protein isolated or integrated. For this application the radiation isotopes are preferably (but not limited to) ^{3}H, ^{35}S, ^{33}P, ^{32}P, ^{14}C and ^{125}I. The invention can also be used for real time product quality control. It will be appreciated that the ASID described above, although intended specifically for use in imaging applications, is not limited to use in such applications. It could be used in other applications. Examples of alternative applications are as part of a radiation impact position detector, as part of a radiation counter, etc.

Thus, there has been described a real time imaging system comprising:-

(a) An active semiconductor pixel imaging device (ASID) consisting of a semiconductor substrate with active circuit elements built on it. The active circuit elements comprise integrated electronic structures such as capacitors and transistors which are able to store the charge created in the semiconductor substrate. The pixel size can be as small as 10µm and the substrate thickness from 200µm to more than 1mm.

(b) Readout and discrimination control electronics not attached to the semiconductor substrate that include CPU and control modules, analogue to digital converters and data reduction processors. These electronics control the readout and processing of the electric signals and are able to address each pixel individually (e.g., by scanning).

(c) An image processor interfaced to the second level electronics that converts the charge density to images.

(d) A workstation or PC that displays the images and carries out image analysis.

(e) A source of radiation located behind the organic or inorganic

object under observation or injected in the form of isotopes to the object under observation and the isotopes consequently attach to selected areas that are imaged. The source of radiation can be X-ray tubes, synchrotron X-rays, 57Co, 60Co, 241Am, 99Tc, 3H, 14C, 35S, 33P, 32P.

There has also been described a method for performing real time imaging comprising the following steps:-

(a) Select an organic or inorganic object to be observed.

(b) Select a radiation source that produces X-rays, γ-rays, β-rays or α-rays and placing the same behind the object under observation or injecting into the object and so that it becomes attached to selected areas.

(c) Arrange that the object under observation either absorbs partially the radiation that was incident upon it and lets through the rest or arranging that selected areas of the object to which the radiation source is attached emit radiation.

(d) Arrange that unabsorbed radiation or radiation that is emitted from selected areas of the object is detected at semiconductor imaging plane or planes that are positioned as close as possible to the object. The imaging plane(s) comprise active pixel semiconductor imaging devices (ASIDs) that convert incident photons or charged radiation to electrical signals stored on active electronic elements built onto pixel cells of the ASIDs.

(e) Individually address by means of control electronics each pixel to read out the stored charge and processed the same without ambiguities as to the point of incidence of the radiation. The stored charge of each pixel is digitized to provide a value representative of the charge imparted to the pixel cell by the incident radiation.

(f) Interface the control electronics to an image processor that receives the information of the digitized charge value and stores it along with the pixel positions to an array. This is done for each pixel.

(g) Find in the image processor the maximum and minimum charge stored and automatically assigning grey or colour scale values to all selected pixels according to their charge and the minimum and maximum grey and colour scale values. This is done according to the formula:

$$\text{Grey scale value of pixel } i = Min_{grey} + \frac{(i_{charge} - Min_{charge})}{(Max_{charge} - Min_{charge})} x (Max_{grey} - Min_{grey})$$

(h) Display each pixel position with a grey or colour scale value as shown above, on a computer screen.

(i) Continuously supply the image processor with new digitized charge values such that a previously stored image array is updated by adding the new charge values for a pixel to the existing value stored for that pixel and displaying the updated image on a display thus providing real time imaging.

(j) Store, at any given instant, the image array and retrieving the same later on to be analyzed, transferred to a different site etc. From a keyboard or using a "mouse" of the image processor, a user can select part of the image corresponding to part of the image array. The image processor the selects the corresponding pixels and displays them on the whole screen (zooming) while automatically adjusting the contrast and resolution according to step (g).

There has also been described a method for discriminating scattered radiation and improving imaging resolution comprising the following steps:-

(a) Receive and record the electrical signal of incoming radiation using an active pixel semiconductor imaging device.

(b) Process stored charge of every pixel including digitizing the same.

(c) Compare the digitized value of every pixel to a reference value and if it is found to be smaller, the pixel value is reset and is not read out to an image processor.

There has also been described a method for automatic design optimization of the imaging plane parameters and automatic choice of the correct design at execution time comprising the following steps:-

(a) Provide a database with materials, compounds, radiation type, energy and desired sensitivity. For each of these items a value for the optimal size of the active semiconductor pixels and a value of th thickness of the active semiconductor pixels is provided.

(b) Use these parameters to determine the optimal design solution by maximizing signal to noise ratio and minimizing cost and complexity.

(c) Alternatively or in addition, use these values automatically to determine a correct imaging plane from a series of planes that are all compatible to control electronics and an image processor of an imaging system during execution.

An ASID and the methods described above can find application in a wide range of applications, including mammography (X-rays 10keV-50keV), tomography (X-rays 20keV-100keV), γ cameras (X-rays > 100keV), X-ray product quality control (20keV-200keV), DNA sequencing, hybridization in situ and hybridization on DNA isolated or integrated with β-ray isotopes 3H (18keV), 14C (155keV), 35S (170keV), 33P (250keV), 32P (1700keV).

CLAIMS

1. An imaging device comprising a semiconductor substrate including an array of pixel cells, each pixel cell comprising an individually addressable active charge storage circuit integral to said semiconductor substrate for accumulating charge resulting from radiation incident on said pixel cell.

2. An imaging device according to claim 1, wherein said active charge storage circuit comprises an integrated charge storage device for accumulating charge.

3. An imaging device according to claim 2, wherein said charge storage device comprises a capacitor and/or a transistor.

4. An imaging device according to claim 2 or claim 3, wherein each pixel cell active charge storage circuit comprises circuitry for selectively resetting said charge storage device, for example after readout of any charge stored thereon.

5. An imaging device according to any one of the preceding claims, wherein said pixel cell size is approximately 150µm across or less, preferably approximately 50µm across or less and more preferably approximately 10µm across.

6. An imaging device according to any one of the preceding claims, wherein said substrate is between 200µm and 1mm thick.

7. An imaging device according to any one of the preceding claims in combination with control electronics including addressing logic for addressing individual pixel cell active charge storage circuits for reading accumulated charge from the active charge storage circuit of a selected pixel cell.

8. An imaging device according to claim 7, wherein said control electronics includes an analogue to digital converter for converting charge read from a said active charge storage circuit into a digital

charge value.

9. An imaging device according to claim 7 or claim 8, wherein at least part of said control electronics is integrated into said semiconductor substrate.

10. An imaging system comprising an imaging device according to claim 8 or claim 9, said imaging system comprising an image processor connected to said control electronics for processing said digital charge values from respective active circuits to form an image for display on a display device.

11. An imaging system according to claim 10, wherein said processor determines maximum and minimum charge values for pixels for display, assigns extreme grey scale or colour values to said maximum and minimum charge values and allocates grey scale or colour values to an individual pixel according to a sliding scale between said extreme values in dependence upon the charge value for said pixel.

12. An imaging system according to claim 11, wherein the grey scale or colour values are allocated in accordance with the following formula:

$$\text{Grey scale value of pixel } i = Min_{grey} + \frac{(i_{charge} - Min_{charge})}{(Max_{charge} - Min_{charge})} x (Max_{grey} - Min_{grey})$$

13. A method for imaging accumulated values corresponding to respective pixel positions within a pixel array, such as, for example, charge values accumulated for respective pixel positions of an imaging device as defined in any one claims 1 to 9, said method comprising:

- determining maximum and minimum accumulated values for pixels within an area of the pixel array to be imaged;
- assigning grey scale or colour values at extremes of a grey or colour scale to be imaged to said maximum and minimum accumulated values; and
- assigning grey scale or colour values to said accumulated

values for individual pixels scaled in accordance with said extreme values; and

- imaging the assigned grey scale or colour values at respective image pixel positions.

14. A method of automatically optimising imaging using, for example, an imaging system according to any one of claims 10 to 12 for different imaging applications where incident radiation leaves a different electrical signal in a pixel cell of a semiconductor substrate dependent on a semiconductor material or compound used and an energy and a type of incident radiation, said method comprising:

- determining an expected best resolution using a centre of gravity technique;

- determining an expected efficiency as a function of radiation type and energy; and

- determining a pixel size and thickness as a function of a selected radiation type and energy and a selected semiconductor material or compound.

15. A method according to claim 14 comprising automatically selecting an imaging device having the determined pixel size and thickness.

16. A method for automatically detecting and eliminating detected pixel value representative of radiation incident on a pixel cell of an imaging device, for example an imaging device according to any one of claims 1 to 9, said method comprising:

- comparing the detected pixel value to a threshold value related to a minimum detected charge value expected for directly incident radiation; and

- discarding detected pixel values less than said threshold value.

17. A method for performing real time imaging of an organic or inorganic object, said method comprising:

- irradiating the object using a radiation source that produces X-rays, γ-rays, β-rays or α-rays;

- detecting at a semiconductor imaging plane or planes of an imaging

device according to any one of claims 1 to 9 unabsorbed radiation or radiation that is emitted from selected areas of the object, whereby charge resulting from incident radiation at respective pixel cells of said imaging device is accumulated in respective active circuits of said pixel cells;

- addressing said active circuits of said pixel cell individually for reading out accumulated charge;
- processing said read out charge to provide image pixel data; and
- displaying said image pixel data.

18. Use of an imaging device according to any one of claims 1 to 9 or of an imaging system according to any one of claims 10 to 12 for chest X-rays, for X-ray mammography, for X-ray tomography, for computerized tomography, for γ-ray nuclear radiography, for β-ray imaging using isotopes for DNA, RNA and protein sequencing, hybridization in situ, hybridization of DNA, RNA and protein isolated or integrated or for X-ray imaging for product quality control.

19. An imaging device substantially as hereinbefore described with reference to the accompanying drawings.

20. An imaging system substantially as hereinbefore described with reference to the accompanying drawings.

21. A imaging method substantially as hereinbefore described with reference to the accompanying drawings.

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Patents Act 1977 Examiner's report to the Comptroller under Section 17 (The Search report)	Application number GB 9410973.3
Relevant Technical Fields (i) UK Cl (Ed.M) H1K (KECCB, KECCX) H4F (FCC) (ii) Int Cl (Ed.5) H01L	Search Examiner R C Hradsky Date of completion of Search 21 September 1994
Databases (see below) (i) UK Patent Office collections of GB, EP, WO and US patent specifications. (ii) ONLINE DATABASES : WPI	Documents considered relevant following a search in respect of Claims :- 1-12, 17, 18

Categories of documents

X: Document indicating lack of novelty or of inventive step.

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A: Document indicating technological background and/or state of the art.

P: Document published on or after the declared priority date but before the filing date of the present application.

E: Patent document published on or after, but with priority date earlier than, the filing date of the present application.

&: Member of the same patent family; corresponding document.

Category	Identity of document and relevant passages	Relevant to claim(s)
X	GB 2262383 A (SONY) whole document	1,17,18
X	GB 2249430 A (MITSUBISHI) whole document	1,17,18

Databases: The UK Patent Office database comprises classified collections of GB, EP, WO and US patent specifications as outlined periodically in the Official Journal (Patents). The on-line databases considered for search are also listed periodically in the Official Journal (Patents).

The Patent Office

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Application No: GB 9703345.0
Claims searched: all

Examiner: Martyn Dixon
Date of search: 21 March 1997

Patents Act 1977
Search Report under Section 17

Databases searched:

UK Patent Office collections, including GB, EP, WO & US patent specifications, in:

UK Cl (Ed.O): H1D (DAB2,DAB7); H4T (TATX)

Int Cl (Ed.6): H01J (31/50,31/52,31/56)

Other: online: WPI, JAPIO

Documents considered to be relevant:

Category	Identity of document and relevant passage		Relevant to claims
X	GB 2228615 A	(Hamamatsu) the whole document	1-7,12,
X	GB 2160013 A	(Hamamatsu) see e.g. fig 9	1-5,7,12
X	GB 1016930 A	(UKAEA) see especially figs 1 and 3	1-7,12
X	EP 0568376 A	(Hamamatsu) see fig 5	1-7,12
X	US 4467189 A	(Hamamatsu) the whole document	1-5,7,12

X	Document indicating lack of novelty or inventive step	A	Document indicating technological background and/or state of the art.
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